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FINITE AMPLITUDE DISTORTION-BASED  
INHOMOGENEOUS PULSE ECHO ULTRASONIC IMAGING

5 BACKGROUND OF THE INVENTION

This invention generally relates to ultrasonic pulse echo imaging, and more specifically, to ultrasonic pulse echo imaging based on the distortion of ultrasonic signals transmitted into samples.

Ultrasonic pulse echo imaging is widely used in many medical applications. While this technique has received wide acceptance, it would be desirable to improve the resolution of the images formed from this technique. For example, ultrasonic pulse echo imaging in inhomogeneous media suffers from significant lateral and contrast resolution losses due to the defocusing effects of the inhomogeneities of the media. The losses in lateral and contrast resolution are associated with increases in the width of the main beams and increases in side lobe levels, respectively.

These two forms of resolution loss represent a significant hurdle to improving the clinical utility of biomedical ultrasonic imaging. A number of research efforts are currently underway to investigate the defocusing effects of tissue and to consider corrective measures. These efforts, however, generally assume linear propagation and base the image-formation process on the reception of the transmitted pulse.

1 SUMMARY OF THE INVENTION

An object of this invention is to improve ultrasonic imaging and methods.

5 Another object of the present invention is to improve the resolution of images formed from ultrasonic echo signals.

10 A further object of this invention is to utilize the enhanced inhomogeneous focusing properties of the finite amplitude distortion generated higher harmonics of an ultrasonic imaging beam in order to obtain improved contrast resolution and lateral resolution images.

15 These and other objectives are obtained with a method and system for imaging a sample. The method includes the steps of generating an ultrasonic signal, directing the signal into a sample, which signal is distorted and contains a first order and higher order component signals at first and higher frequencies respectively. The received distorted signal is  
20 processed, and an image is formed, and then displayed, from one of the higher order component signals of the received distorted signal.

25 With the preferred embodiment of the invention disclosed herein in detail, the ultrasonic image is based on one of the received finite amplitude distortion component (or nonlinearly-generated higher harmonics) associated with the transmitted signal. In the simplest case, in which the transducer emits negligible energy in the second harmonic bandwidth,  
30 such an image can be formed by adding an initial high pass filtering of the received signal. In general,

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1 such an image can be formed by using a two pulse  
transmit, receive, normalize, and then high pass  
filtering scheme. Such a two pulse scheme can be used  
5 and higher harmonic bandwidths.

Further benefits and advantages of the  
invention will become apparent from a consideration of  
the following detailed description given with  
reference to the accompanying drawings, which specify  
and show preferred embodiments of the invention.

#### BRIEF DESCRIPTION OF THE DRAWINGS

Figure 1 shows an ultrasonic imaging system  
embodying the present invention.

Figures 2a, 2b and 2c show various  
15 parameters associated with the linear propagation  
results for a focused 2 MHz Gaussian Transducer  
operating in a liver medium.

Figure 3 shows discrete harmonic velocities  
used to compute various values associated with a 2 MHz  
20 propagation in a liver medium.

Figures 4a and 4b display nonlinear  
propagation results for a focused 2 MHz Gaussian  
transducer.

Figure 5a shows the log-scaled, normalized  
25 one-way focal plane profiles of the 2 MHz fundamental,  
the 4 MHz second harmonic, and the 4 MHz fundamental,  
and Figure 5b shows the corresponding two way profiles  
for these beam patterns.

Figure 6a shows the on-axis source plane and  
30 the subsequent focal plane for a 2 MHz Gaussian  
source.

1           Figure 6b illustrates the corresponding  
spectrums of the source and focal planes shown in  
Figure 6a.

5           Figure 6c shows the focal nonlinear  
distortion pulse obtained by constructing the waveform  
using only the focal spectral information shown in  
Figure 6b from 3 to 8 MHz.

10           Figure 7 shows the focal plane profile from  
a 2 MHz continuous wave propagation and a 2 MHz  
centered pulse propagation.

          Figure 8 is a table giving the on-axis  
source plane intensity versus the received second and  
third harmonic levels and focal field parameters.

15           Figure 9 is a Table providing focal and  
received second harmonic levels versus focal length.

20           Figures 10a and 10b show the log-scaled  
first, second and third harmonic axial amplitudes for  
the focused 2 MHz Gaussian transducer and the  
corresponding log-scaled focal plane radial beam  
profiles.

          Figure 11 illustrates scaled axial  
amplitudes for a 4 MHz second harmonic and 4 MHz  
fundamental harmonic beams in a liver medium.

25           Figures 12a-12f depict pairs of one-way  
focal plane harmonic amplitude diameters and the  
corresponding average radii for an unjittered, or  
homogeneous, path and an abdominal wall-jittered  
propagation path.

30           Figures 13a-13d show two-way average radii  
results for the abdominal wall-jittered propagation  
path represented in Figures 12a-12f.

1           Figures 14a and 14b display normalized two-  
way averaged radial results from five abdominal wall-  
jittered propagations for the 2 MHz fundamentals, the  
4 MHz second harmonics and the 4 MHz fundamentals, and  
5   the corresponding radially integrated magnitudes.

          Figures 15a and 15b show normalized two-way  
averaged radial results from five abdominal wall-  
jittered propagations for the 4 MHz fundamentals, the  
8 MHz second harmonics and the 8 MHz fundamentals, and  
10   the corresponding radially integrated normalized  
magnitudes.

          Figures 16a and 16b show normalized two-way  
averaged radial results from five breast jittered  
propagations for the 2 MHz fundamentals, the 4 MHz  
15   second harmonics and the 4 MHz fundamentals, and the  
corresponding radially integrated magnitudes.

          Figure 17 is a table giving the -20dB full-  
widths for the average two-way profiles shown in  
Figures 14a, 15a and 16a and the full-widths at the  
20   0.9 level for the integrated profiles of Figures 14b,  
15b and 16b. Also shown are the corresponding results  
from the 8 MHz breast jittered propagations.

          Figures 18a-18d show an imperfect source  
pulse, linearly-scaled and log-scaled focal spectrums  
25   of that pulse, and the corresponding nonlinear  
distortion pulse obtained by constructing with the  
spectral information in Figure 18c.

          Figure 19a shows the log-scaled focal  
spectrum of Figure 19c overlaid with the focal  
30

1 spectrum from the same source using a half amplitude  
version of the source pulse, as depicted in Figure  
19c.

5 Figure 19b shows the resulting difference  
spectrum computed for the two spectrum shown in Figure  
20a.

10 Figure 19c shows the corresponding nonlinear  
distortion pulse obtained by constructing with the  
spectral information in Figure 19b and starting at  
2.75 MHz.

#### DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENT

15 Figure 1 illustrates ultrasonic imaging  
system 10. A pulse generator 12 and a function  
generator 14 produce a sinusoidal pulse ultrasonic  
signal of, for example, 2.0 MHz at a pulse repeat  
frequency of, for instance, 1 kHz. This signal is  
sent to amplifier 20, which amplifies the signal and  
transmits the amplified signal to transducer-receiver  
unit 22, and this unit then transmits the signal into  
sample 24.

20 In this sample 24, the input signal is both  
distorted and reflected. The distortion creates a  
distorted signal having a multitude of component  
signals, each of which has a respective frequency or  
25 frequency bandwidth. The distorted signal is  
reflected by sample 24, and this reflected signal is  
received by transducer-receiver unit 22, amplified by  
pre-amplifier 30, and then further amplified by  
amplifier 32. The received and amplified signal is  
30 then sent through a high-pass filter 34 to enhance the  
relative strength of the desired higher harmonic

1 component of the received signal. The resulting  
signal is digitized in analog-to-digital converter 36,  
and then processed by processor 40 to produce an  
image.

5 This image may be displayed on a video monitor  
42, stored on a video cassette recorder (VCR) 44,  
output on a printer device 46, or stored in any of a  
variety of hard copy storage devices 50, such as  
10 medical film recorders, digital tape machines, optical  
disks, magnetic tapes and disks or the like. Suitable  
means may be used to move the focal point of the  
transmitted signal around sample 24. For instance,  
transducer-receiver unit 22 may be a phased array unit  
15 having electrical circuitry to move the focal point of  
the transmitted signal. Alternatively, a motor 52 may  
be employed to move transducer-receiver unit 22 and  
thereby move the focal point of the transmitted signal  
around sample 24.

20 System 10 can also be operated in a two  
pulse scheme or mode. In this mode of operation,  
system 10 generates and transmits into sample 24 two  
different pulses. Preferably, the transmitted signals  
are identical except that one of them is scaled up in  
pressure. The pulses are transmitted one after  
25 another with, for example, approximately 1/4000 second  
interval between them. The reflected, distorted  
signals from both pulses are received by transducer  
unit 22, and these signals are then digitized in  
analog-to-digital converter 36. The digital data  
30 values obtained from the first pulse are stored,  
scaled and then subtracted from the digital data

1 values obtained from the second pulse, producing a  
difference or resultant signal. Subsequent processing  
of this difference signal in system 10 is the same as  
in the above-described one pulse case. One suitable  
5 procedure for scaling the digital data values obtained  
from the first pulse is discussed below.

10 Several computational models exist which can  
accurately describe the finite amplitude propagation  
of a continuous beam. Such models can be extended to  
compute the case of a propagating pulse as well.  
These models account for the effects of diffraction,  
nonlinearity and absorption. One of these models,  
referred to as the NLP model is described in "New  
approaches to nonlinear diffractive field propagation"  
15 J. Acoust. Soc. Am. 90, 488-499 (1991) by P.T.  
Christopher and K.J. Parker, the disclosure of which  
is herein incorporated by reference, and this model  
was used to compute the following linear and nonlinear  
examples. This model has been updated to account for  
20 the effects of dispersion. The associated harmonic  
velocities were computed using an algorithm disclosed  
in "Modeling acoustic field propagation for medical  
devices," Ph.D. thesis, U. of Rochester (1993) by T.  
Christopher, the disclosure of which is herein  
25 incorporated by reference.

The propagations discussed below are for a  
Gaussian apodized, axially symmetric focused source.  
This form of device offers excellent image quality and  
produces a field comparable to that of the two  
30 dimensional array-based transducers now being  
developed.



1 Many biomedical imaging devices are not  
axially symmetric, though. The finite amplitude beams  
produced by such devices are well described by the  
computations for the axially symmetric transducers  
5 discussed here in detail, though. In measuring the  
nonlinear harmonic generation from an unfocused  
rectangular source, Kamakura, Tani, Kumamoto and Ueda  
noted in "Harmonic generation in finite amplitude  
sound beams from a rectangular aperture source," J.  
10 Acoust. Soc. Am. 97, 3510-3517 (1995), that "the  
[nonlinear] harmonic pressure levels in the far field  
[were] almost the same as from a circular aperture  
source with equal face area and equal initial  
pressure, independent of the source levels." Though  
15 this result was obtained for only one device (with a  
ratio of source side lengths or aspect ratio of 11 to  
6), the higher harmonic pressure levels associated  
with a non-axially symmetric device are approximately  
equal to those of the corresponding axially symmetric  
20 source. More importantly, the lack of axial symmetry  
does not affect the relative sidelobe advantages  
exhibited by the nonlinearly-generated harmonics in a  
homogeneous propagation. These homogeneous path  
sidelobe level advantages are the basis for the  
25 imaging-relevant advantages of the higher harmonics in  
an inhomogeneous propagation.

Figures 2a, 2b and 2c show various  
parameters associated with the linear propagation  
results for a focused 2 MHz Gaussian transducer  
operating in a liver medium. In particular, the  
30 source plane amplitude profile, the on-axis amplitude,

and the radial focal plane (at  $Z = 6\text{cm}$ ) beam profile  
are shown in Figures 2a, 2b and 2c respectively. The  
results shown in Figures 2a, 2b and 2c were obtained  
by computing the linear, liver path propagation of the  
field of a focused 2 MHz Gaussian source using the NLP  
beam propagation model. The NLP model propagates a  
planar, normal velocity description of the acoustic  
field. No inhomogeneities or phase aberrations were  
accounted for in this propagation or any of the  
subsequent propagations considered immediately below.  
The relevant liver propagation parameters used were  $c$   
 $= 1570\text{ m/s}$ ,  $\rho = 1.05\text{ g/cm}^3$ ,  $\alpha = 0.03\text{ Np/cm}$  and  $b = 1.3$   
(where  $\alpha$  and  $b$  are the coefficients describing  
absorption in a power law form).

The Gaussian shading of the magnitude of the  
source plane normal velocity field was such that the  
half-amplitude radial distance was  $0.84\text{ cm}$ . The on-  
axis, source plane RMS acoustic intensity ( $\rho c |u|^2/2$ ,  
where  $u$  is the acoustic particle velocity) for the  
field was  $2\text{ W/cm}^2$ . The radial extent of the source  
was  $1.5\text{ cm}$ . The source plane field was focused using  
a spherically-focusing factor ( $e^{j\theta(r)}$ , where  $\theta(r) =$   
 $(2\pi f/c)\sqrt{r^2+F^2}$ ). The geometric focal length  $F$  was  $6\text{ cm}$   
and the sound speed ( $c$ ) used to compute  $\theta(r)$  was that  
of water ( $1500\text{ m/s}$ ).

Figures 2a and 2b depict the normal velocity  
magnitudes of the Gaussian transducer's source plane  
and on-axis fields respectively. Figure 2c displays  
the focal plane ( $z = 6\text{ cm}$ ) radial profile of the 2 MHz  
field. The drop in the magnitude of the field from  
the mainlobe to the first sidelobe in Figure 2c is  $36\text{ dB}$ .  
In the absence of strong medium phase aberration

this allows the device to produce high contrast  
1 images.

The same 2 MHz Gaussian source was then  
propagated nonlinearly through the same liver path.  
The nonlinear parameter  $\beta$  used to represent liver was  
5 4.7. The NLP model used 4 harmonics (2, 4, 6 and 8  
MHZ) to compute the pre-focal region ( $z = 0$  to  $z = 3$   
cm) propagation and up to 10 harmonics to represent  
the subsequent focal and post-focal region  
propagation. The harmonic velocities were computed  
10 using the algorithm described in "Modeling acoustic  
field propagation for medical devices," T.  
Christopher, Ph.D. Thesis, University of Rochester  
(1993). The fundamental or 2 MHz component had a  
propagation speed of 0.157 cm/microsecond (given above  
15 as  $c$ ). The discrete harmonic velocities used by NLP  
to compute the 2 MHz propagation are shown in Figure  
3.

Figure 4a displays the axial magnitudes of  
the fundamental, second harmonic, and third harmonic  
20 fields, at 102a, 102b and 102c respectively, as  
computed for the nonlinear propagation. The  
fundamental or 2 MHz axial curve is only slightly  
different from the corresponding linear curve shown in  
Figure 3b. At  $z = 6$  cm the nonlinear 2 MHz curve is  
25 about 1% lower than the 2 MHz linear curve. This  
difference was due to growth of the higher harmonics  
in the nonlinear propagation. In Figure 4b the  
corresponding focal ( $z = 6$  cm) pressure waveforms from  
the linear and nonlinear computations are displayed at  
30 104a and 104b respectively. The pressure waveforms  
were obtained by converting NLP's normal velocity

output to pressure using the impedance relation (all  
subsequent pressure waveforms were obtained in this  
way).

The 2 MHZ or fundamental beam pattern 104b  
associated with the nonlinear propagation is almost  
identical with the 2 MHZ beam pattern 104a of the  
linear propagation. Only in a linearly scaled overlay  
plot of the two beam patterns are there visible  
differences. These differences are very small and are  
limited to the near axis portions of the beam  
patterns. Only at much higher source intensities are  
the effects of nonlinearity significant to the details  
of the fundamental's field. These results are  
consistent with the empirical observation that linear  
modeling of biomedical ultrasonic device fields  
accurately describes their (linear-based homogeneous  
path) imaging performance.

Figure 5a depicts the 2 MHZ fundamental and  
4 MHZ second harmonic focal plane beam amplitude  
profiles at 106a and 106b. Also shown in Figure 5a at  
106c is the corresponding 4 MHZ fundamental profile.  
The 4 MHZ fundamental result was obtained by computing  
the linear propagation of the same Gaussian transducer  
operating at a source frequency of 4 MHZ. All three  
beam profiles in Figure 5a have been normalized to  
have on-axis field magnitudes of 1. The finite  
amplitude distortion-generated second harmonic focal  
profile 106b has a slightly broader mainlobe than the  
corresponding (4 MHZ) fundamental profile 106c. The  
radial half-amplitude distance of the second harmonic  
profile is 36% greater than that of the 4 MHZ  
fundamental profile (0.0983 cm versus 0.0723 cm). The

second harmonic profile also has much lower sidelobes  
1 than the 4 MHz fundamental profile.

For imaging purposes, the two-way focal  
plane beam pattern of the Gaussian transducer is of  
interest. The two-way focal beam pattern accounts for  
5 both the characteristics of the transmitted pulse in  
the focal plane and the corresponding sensitivity of  
the transducer to pulses reflected back from this  
plane. For linear propagations, the two-way beam  
pattern for a given depth can be obtained by squaring  
10 the corresponding transmit or one-way beam pattern.  
In Figure 5b the normalized two-way linear beam  
patterns for the Gaussian transducer operating at 2  
and 4 MHz are depicted at 110a and 110b. These curves  
were obtained by squaring the corresponding one-way or  
15 transmit beam patterns shown in Figure 5a.

Also shown in Figure 5b at 110c, is the two-  
way focal plane beam pattern associated with the 4 MHz  
second harmonic field. Since the amplitudes of the  
reflected pulses are much smaller than the transmitted  
20 pulses, the propagation of the reflected field back to  
the transducer is essentially linear. Thus, the two-  
way focal plane beam pattern shown at 110b for the  
second harmonic was obtained by multiplying the  
corresponding one-way pattern 106b shown in Figure 5a  
25 by the 4 MHz fundamental one-way pattern 106b also  
shown in Figure 5a.

The second harmonic's two-way beam pattern  
has a half-amplitude mainlobe width (or -6 dB  
beamwidth) that is 12% greater than that of the  
30 corresponding beam pattern 106c. The -20 dB beamwidth  
of the second harmonic is 13% greater than that of the

4 MHZ fundamental. The sidelobe advantage displayed  
1 in the focal plane profiles of Figure 5a is maintained  
in the two-way results shown in Figure 5b.

These homogeneous results show that the  
second harmonic field of a focused, apodized  
5 transducer offers advantages in contrast resolution  
over the corresponding linear transducer field.

A pulse propagation was next considered for  
the 2 MHZ Gaussian source. The on-axis, source plane  
pressure pulse used is displayed in Figure 6a as the  
10 desired curve 112a. This pulse was computed by  
applying a Gaussian window to a 2 MHZ cosine. The  
peak pressure of the pulse was the same as for the 2  
W/cm<sup>2</sup> continuous case considered above. The radial  
amplitude shading or apodization and the spherical  
15 focusing of the source field were also the same as in  
the previous continuous wave case. The initial source  
plane pulse consisted of 128 samples across 8  
microseconds.

The magnitude of the Fourier transform of  
the 8 microsecond long source pulse is shown at 114a  
20 in Figure 6b. A straightforward implementation of the  
nonlinear imaging system and method requires  
negligible overlap between the sources's spectral  
bandwidth and that of the nonlinear second harmonic  
25 (more generally, this would also insure negligible  
overlap between any of the successive harmonic  
spectral bands). The source spectrum 114b shown in  
Figure 6b meets this requirement. Nonlinear images  
based on source pulses with broader spectrums or with  
30 significantly more energy in the second harmonic  
bandwidth than the one depicted at 112b and 114b in

Figures 6a and 6b could be obtained by using an  
1 alternative nonlinear imaging scheme described below.

The source plane was then defined using the  
64 harmonic Fourier transform of the source pulses.  
This multiharmonic source radius was then input into a  
5 pulse-propagating version of the NLP model (a model  
similar to the lithotripter model presented in  
"Modeling the Dornier HM3 Lithotripter." T.  
Christopher, J. Acoust. Soc. Am. 3088 - 3095 (1994).  
The focal output of the resulting nonlinear pulse  
10 propagation is also shown in Figures 6a and 6b as  
solid curves. The focal pulse waveform has slightly  
smaller peak positive and negative pressures than the  
corresponding continuous waveform shown in Figure 4b.  
Consistent with the smaller amplitudes, the focal  
15 pulse is also less distorted than the continuous  
waveform. The ratio of the second harmonic's focal  
amplitude to that of the fundamental's for this pulse  
propagation was 70% of the same ratio for the  
corresponding continuous source considered above.

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25 In Figure 6c, the waveform associated with  
only the spectral bands of the second, third, and the  
first half of the fourth harmonic (3 through 8 MHz) is  
shown. This waveform was computed as a high pass  
filtered reconstruction of the spectral information  
depicted in Figure 6b. A rectangular window with a  
transition at 3 MHz was used in filtering the  
transform data. The straightforward nonlinear imaging  
approach disclosed herein may use a distortion pulse  
like that shown in Figure 6c in order to image the  
30 scattering medium.

Not shown for the pulse propagation

considered are the axial and radial harmonic  
1 descriptions. These descriptions were found to be  
identical in form to those computed for the nonlinear  
propagation of the continuous 2 MHz field. All other  
Gaussian-windowed cosine pulses were found to produce  
5 identical axial and radial harmonic patterns. The  
temporal peak amplitude profile of the high pass  
filtered-focal plane data also very closely followed  
the form of the 4 MHz second harmonic's amplitude  
profile. The temporal peak amplitude profile for the  
10 pulse propagation considered herein is shown at 116a  
in Figure 7. This profile was computed using the  
focal plane frequency data in the range of 3 to 8 MHz  
(like the waveform in Figure 6c). Also shown at 116b  
and 116c in Figure 7 are the corresponding 4 MHz  
15 second harmonic profile and a second harmonic bandpass  
filtered (3 to 5 MHz)-temporal peak amplitude profile  
for the same pulse propagation. The similarity of the  
two peak amplitude profiles suggests that there is  
very little energy above 5 MHz for this pulse  
20 propagation. The nonlinear harmonic beam and peak  
amplitude patterns produced by any reasonable source  
pulse can be computed using an appropriate continuous  
approximation of the source.

In order for the nonlinearly-generated  
25 higher harmonics to be available for imaging in an  
inhomogeneous media, the received higher harmonics  
amplitudes cannot be too low relative to the received  
linear (or transmitted) signal and the transducer's  
dynamic range. Also, to be available for imaging use,  
30 preferably the associated in vivo field amplitudes  
have a mechanical index less than 1.9.



For the previously considered nonlinear 2  
1 MHZ, liver-path propagation, the amplitudes at the  
focus ( $z = 6$  cm) of the fundamental and second  
harmonic were 0.943 MPa and 0.166 MPa, respectively.  
The second harmonic amplitude was thus 15.09 dB below  
5 the amplitude of the 2 MHZ transmitted wave. As a  
wave reflected at the focal point travels the 6 cm  
back to the transducer/receiver, frequency dependent  
attenuation reduces the second harmonic by an  
additional 5.62 dB relative to the 2 MHZ component.  
10 The received 4 MHZ component would thus be 20.71 dB  
below the transmitted 2 MHZ component. The  
corresponding figures for the 6 MHZ third harmonic are  
27 dB down at the focus and 39.23 dB down for received  
signals. This calculation, it should be noted, does  
15 not include the effect of the possible additional  
relative weakening of the second harmonic received  
signal due to coherent reflectors at the focus in  
combination with the smaller mainlobe of the second  
harmonic. This effect is not significant to many bio-  
20 ultrasound imaging applications, though.

Table 1 of Figure 8 displays the second and  
third harmonic received levels at on-axis, source  
plane intensity values of 0.5, 1, 2, 4 and 8 W/cm<sup>2</sup>.  
For a given source plane intensity, pulsed devices  
25 would have slightly larger received level  
differentials than those shown in Table 1. For the  
pulsed propagation considered above, this additional  
gap would be 0.86 dB for the second harmonic (based on  
second harmonic bandpass reconstruction and a  
30 comparison of the received peak positive pressure  
levels).

Current biomedical ultrasonic imaging  
1 transducers have dynamic ranges of about 100 dB. Even  
with decreased sensitivity above the transmit  
frequencies, these devices are capable of creating  
second harmonic images. This capability has been  
5 demonstrated by the creation of second harmonic  
contrast agent-response images. Alternatively, a  
separate receiver device with appropriate frequency  
response in the desired nonlinear distortion bandwidth  
can be used.

10 The effect of focal length on the received  
second harmonic levels for this Gaussian transducer  
operating at 2 W/cm<sup>2</sup> is shown in Table 2 of Figure 9.  
From a focal length of 4 cm up to a focal length of 12  
cm the received second harmonic levels dropped off by  
15 7.32 db. Also shown in Table 2 are the corresponding  
focal second harmonic levels. These levels remained  
very constant and thus revealed the decreases in  
received levels as almost entirely due to increased  
return trip distances.

20 The peak positive and negative pressures of  
the in vivo nonlinear waveform shown in Figure 4b were  
1.12 and -0.84 MPa, respectively. The -0.84 peak  
rarefaction pressure corresponds to a mechanical index  
(MI) of 0.59. The highest preferred level for the  
25 mechanical index is 1.9. Shown in Table 1 are values  
of the computed minimum focal pressure and associated  
mechanical index for this and four other values of  
source plane intensity. The minimum pressures and  
thus mechanical indices given in Table 1 have been  
30 corrected for the effects of nonlinearity. A linear-  
only computation would result in larger negative

1 pressures and MI values, in particular at the highest  
two source intensity levels.

5 The numbers shown in Table 1 show that for  
in vivo propagations similar to the one considered  
here, finite amplitude distortion-based images are  
10 readily obtainable within the current mechanical index  
safety limit. Even at the lowest source intensity  
case considered ( $0.5 \text{ W/cm}^2$ ), a largely second  
harmonic-based image can be obtained by simply  
filtering out the transmitted frequency or  
15 frequencies. Additionally the second and third  
harmonic received levels offer some real-time feedback  
on the magnitude of the focal field amplitudes  
themselves. Finally, the results displayed in Table 2  
suggest that second harmonic imaging may be available  
at a wide range of focal depths.

20 The formation of the higher harmonic  
constituent beams in a propagation finite amplitude  
beam is a continuous process. In the case of the 2  
MHZ Gaussian-shaded, focused beam considered above,  
the 4 MHZ second harmonic, the 6 MHZ third harmonic,  
and additional higher harmonics are continuously and  
cumulatively produced by the beam as it propagates  
away from the source. Of interest here is the  
25 production and focusing of these nonlinear higher  
harmonic beams between the source and focal plane.

30 The origin of the higher harmonic beams is  
the ongoing nonlinear distortion of the propagating  
waves comprising the (total harmonic) focused beam.  
The physical effects of diffraction and absorption  
concurrently act on the higher harmonic beams and thus  
further define their propagation as well as contribute

to changes in the resulting focused beam. The NLP  
1 model assumes that the nonlinear or finite amplitude  
distortion acts in a plane wave fashion on the waves  
comprising the focused beam. The NLP model uses the  
frequency domain solution to Burgers' equation in an  
5 incremental  $\Delta z$  fashion to account for this plane wave  
distortion approximation.

The frequency domain solution to Burgers'

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equation as used in the NLP model can be written

$$\frac{\partial u_n}{\partial z} = j \frac{\beta \pi f}{2c^2} \left( \sum_{k=1}^{n-1} k u_k u_{n-k} + \sum_{k=n}^N n u_k u_{k-n}^* \right), \quad n = 1, 2, \dots, N$$

where  $f$  is the fundamental frequency and  $u_n$  is the  $n$ th term in an  $N$  term complex Fourier series describing the temporal normal velocity waveform at a radial point in the radial description of the field. The first summation term in the parentheses represents the accretion of the  $n$ th harmonic by nonlinear combination of other harmonics that have a sum frequency of  $nf$ . The second summation term (with conjugation) represents the depletion of the  $n$ th harmonic to other harmonics with a difference frequency of  $nf$ . For the case of a Fourier representation of a (periodic) pulse waveform, this accretion and depletion of harmonics results in some interesting phenomenon including the production of a distortion bandwidth below the fundamental bandwidth.

Of interest are the terms in equation (1) contributing to the growth of the second harmonic and, to a lesser extent, the third harmonic. When  $n$  equals 2 in equation (1), the positive

contributions to  $\frac{\partial u_2}{\partial z}$  come from the  $1u_1u_1$  term comprising

the first summation. The negative contributions to the second harmonic are represented by the second summation in equation (1) and for the propagations

under consideration here can be approximated by the  
1 second term in that summation,  $2u_3u_1^*$ . Even this term,  
though, is negligible throughout most of the source to  
focal region propagation due to the relatively small  
amplitudes of the third harmonic. Thus, the nonlinear  
5 production of the second harmonic throughout most of  
the relevant propagation region is simply proportional  
to the square of the amplitude of the fundamental.

When  $n$  equals 3 in equation (1), the  
positive contributions to  $\frac{\partial u_3}{\partial z}$  come from the first two  
10 terms comprising the first summation. These two terms  
sum to  $3u_1u_2$ . The first term of the corresponding  
negative contributions to the third harmonic is  $3u_4u_1^*$ .  
This term and it's successors are negligible for all  
but the focal region of the highest intensity  
15 propagations considered herein.

The harmonic sources of the second harmonic  
and third harmonic are thus  $1u_1u_1$  and  $3u_1u_2$ ,  
respectively. At a given point in the field of the  
propagating beam then, the finite amplitude production  
20 of the second harmonic is proportional to the square  
of the fundamental harmonic's amplitude. The  
production of the second harmonic off the beam axis is  
very small since the amplitude of the fundamental beam  
there is quite small. The third harmonic is produced  
25 in proportion to the product of the first and second  
harmonics and thus its nonlinear production is even  
more strongly weighted towards the beam axis.

Also of interest is how the production of  
the second and third harmonics vary with the  $z$   
30 coordinate. Neglecting the effects of absorption and  
approximating the effects of focusing by assuming

spherically-converging wave propagation, the  
1 amplitudes of the fundamental harmonic's mainbeam  
increases approximately linearly with distance from  
the source to the focus. At  $z = F/2$  the on-axis  
amplitudes of the beam are about twice the  
5 corresponding source plane amplitudes. Following from  
the same approximations, the fundamental's mainbeam  
width at  $z = F/2$  is about half its source plane width.  
Thus, at  $z = F/2$  the area of the fundamental's  
mainbeam is about  $1/4$  the corresponding source plane  
10 area. Together, this relation and the previous  
amplitude relation suggest that the second harmonic  
beam production rate versus  $z$  is constant ( $z < F$ ),  
with rate losses due to diminishing fundamental  
mainbeam area balanced by the concurrent gains due to  
15 increased fundamental amplitudes.

The fact that the third harmonic production  
is proportional to the product of the fundamental and  
second harmonic amplitude, though, implies that the  
production of the third harmonic is strongly weighted  
20 towards the focal region. In Figure 10a the log-  
scaled, axial amplitudes of the fundamental, second,  
and third harmonics are displayed at 120a, 120b and  
120c respectively. The source was the same focused 2  
MHZ Gaussian source considered above. The medium's  
25 parameters were again those of liver. Consistent with  
the above discussion, the amplitudes of the second  
harmonic exhibit a relatively large gain in its growth  
from low near field values to significant focal  
amplitudes. The third harmonic exhibits an even  
30 higher gain, approximately duplicating the growth in  
gain from the fundamental to the second harmonic.

Both harmonics, though, display post-focal region  
1 amplitude declines which parallel those of the  
fundamental.

In Figure 10b the corresponding log-scaled,  
focal plane ( $z = 6$  cm, one way) radial beam profiles  
5 are displayed at 122a, 122b and 122c respectively.  
The fundamental profile drops 49.8 dB over the 1 cm  
radial range displayed. The second harmonic  
approximately squares this decline in dropping 85.3  
dB. The third harmonic then continues the  
10 relationship in dropping 120.5 dB. These declines  
reflect the second and third harmonic, finite  
amplitude production rates discussed above. At  $z = 8$   
cm this relationship between the harmonic beam  
profiles continued to hold.

In Figure 11, the axial amplitudes of the 4  
15 MHZ second harmonic beam, shown at 124a, are overlaid  
with the corresponding 4 MHZ fundamental harmonic  
beam, shown at 124b. The on-axis source of both of  
the respective propagations was  $2 \text{ w/cm}^2$ . The axial  
20 curves have been normalized to be unity at  $z = 6$  cm  
and log-scaled. The 4 MHZ second harmonic focal  
amplitude was originally (pre-normalization) 15.0 db  
before the corresponding fundamental value. The two  
curves are close through the focal region and then  
25 depart shortly after  $z = 6$  cm as the 4 MHZ linear  
curve rapidly declines.

The results displayed in Figures 10a, 10b,  
and 11 show that the 4 MHZ second harmonic beam may be  
less susceptible to the defocusing effects of near  
30 field phase aberrations than a 4 MHZ fundamental beam.  
Since only a fraction of the second harmonic beam



forms in the near field, only this fraction could be  
1 redirected or defocused by near field jitter. The  
corresponding 2 MHz fundamental beam, though, would  
pass in its entirety through the aberration and suffer  
the consequent defocusing effects including increased  
5 focal plane sidelobe levels. Secondly, though,  
these higher 2 MHz sidelobe levels could, in turn,  
increase the off-axis nonlinear production of the 4  
MHz second harmonic.

In order to investigate the effects of  
10 tissue-based phase aberration on the characteristics  
of linear and nonlinear beams, planes of phase delay  
or jitter were introduced into linear and nonlinear  
propagations of the focused Gaussian transducer.  
These phase delay planes were computed using measured  
15 time delays from 5 abdominal wall layers and 5 breast  
wall layers. The 5 abdominal wall specimens had layer  
thicknesses of 2.5, 2.0, 1.5, 1.5, and 1.0-3.0 cm (a  
non-uniform layer) giving an average thickness of 1.9  
cm. The 5 breast wall specimens had layer thicknesses  
20 of 1.5-2.5, 3.0-3.5, 3.5, 4.0 and 2.0-2.5 cm, giving  
an average thickness of 3.0 cm. The average  
thicknesses of the non-uniform layers were used to  
compute the 5-layer averages. All of the measured  
abdominal wall and breast wall layers contained an  
25 outer skin layer.

Each of the measured abdominal wall time  
delay planes was converted to an equivalent 2 MHz  
phase delay plane. Each of these delay planes was  
then scaled by 0.5 (i.e., each phase delay was reduced  
30 by a factor of 2) and then applied twice to a given  
beam propagation to represent the cumulative aspect of

the actual tissue delays. In applying a single delay  
plane, the 2 MHZ phase delay values were scaled for  
appropriate application to each harmonic present in  
the computed field. The first delay plane was  
encountered by the propagating field at  $z = 0.5$  cm and  
the second plane at  $z = 1.5$  cm. Further, subdivision  
c and subsequent applications of the abdominal wall  
delay data did not appear to be necessary since it did  
not significantly change the resulting focal plane  
fields. Thus, the 2 delay plane application scheme  
satisfied the thin lens approximation. The breast  
delay planes were likewise applied in two steps, the  
first at  $z = 1$  cm and the second  $z = 2$  cm. For both  
tissue types, the  $z$  placement was selected to  
represent the average slice thickness and also to be  
convenient for the  $\Delta z$  step size utilized by the linear  
propagation.

Figures 12a-12f depict pairs of one-way  
focal plane harmonic amplitude diameters and the  
corresponding average radii for an unjittered (or  
homogeneous) path and an abdominal wall-jittered  
propagation path. The propagation parameters of both  
mediums were again those of liver. In Figure 12a the  
corresponding focal plane diameters for the unjittered  
and jittered 2 MHZ fields are overlaid at 132a and  
132b respectively. In Figure 12b the corresponding  
average radii are shown at 134a and 134b respectively  
for the 2 MHZ fields. The average radii were obtained  
by averaging the focal plane grid of amplitudes around  
the axis. The corresponding results for the second  
harmonic 4 MHZ are shown at 136a, 136b, 140a and 140b  
in Figures 12c and 12d, and for the fundamental 4 MHZ

fields are shown in at 142a, 142b, 144a, and 144b in  
1 Figures 12e and 12f. In all of these Figures, the  
unjittered field is displayed using solid curves.

In Figures 12a-12f, two basic effects of  
jitter are visible. The first is the increased  
5 sidelobe levels associated with the defocusing of the  
abdominal wall phase delays. This sidelobe effect is  
more prominent for both of the 4 MHZ fields than for  
the 2 MHZ fundamental field. The shorter wavelengths  
of these 4 MHZ fields allows for greater de-focusing  
10 by the phase screen. The second harmonic 4 MHZ field  
also gets a sidelobe level increase from the  
corresponding increase in the 2 MHZ field's sidelobe  
levels.

The second impact revealed in Figures 12a-  
12f is that there are small changes in the mainlobes  
15 of all three harmonic fields. In particular, in the  
diameter Figures 12a, 12c and 12e, a shift in the peak  
or center of the jittered lobes can be seen. In the  
corresponding average Figures 12b, 12d and 12f though,  
20 the impact of the jitter is negligible down to  
approximately 20 dB below the peak on-axis value.  
Thus, the impact of this abdominal wall-jittering does  
not appear to involve any significant broadening of  
the mainlobes. Finally, both of the jittered 4 MHZ  
25 mainlobes show decreases in peak amplitudes which are  
consistent with the increased energy present in the  
sidelobe regions.

Figures 13a-13d show two-way average radii  
results for the same abdominal wall-jittered  
30 propagation. Each of the curves shown was obtained by  
radially averaging the corresponding two-way planar

data and then scaling the on-axis value to unity. The  
1 2 MHz fundamental, 4 MHz second harmonic, and 6 MHz  
third harmonic average radii are shown at 146a, 146b  
and 146c in Figure 13a. The 2, 4 and 6 MHz  
fundamentals are shown at 146a, 150a and 150b in  
5 Figure 13b. The average two-way sidelobe levels of  
the second and third harmonic can be seen to be  
significantly lower than those of any of the  
fundamental harmonics. In Figure 13c, the two 4 MHz  
average two-way profiles are shown at 152a and 152b.  
10 The 4 MHz fundamental curve has a slightly narrower  
mainlobe (9.5% at 20 dB down from peak) and  
significantly higher sidelobe levels than the 4 MHz  
second harmonic profile. Likewise, Figure 13d shows  
the 6 MHz fundamental and 6 MHz third harmonic two-way  
15 profiles at 154a and 154b. The 6 MHz fundamental  
curve is 10.9% narrower at the -20 dB level and also  
shows higher sidelobe levels than the third harmonic  
curve.

The results shown in Figures 13a-13d show  
20 that the second and third harmonics maintain lower  
sidelobe levels than the corresponding fundamental  
harmonics in propagating through abdominal wall. In  
Figures 14a and 14b, radial results obtained by  
averaging across 5 abdominal wall-jittered  
25 propagations are shown. In Figure 14a the normalized  
radial averages from the 5 two-way planar amplitude  
data sets are depicted. The average 2 MHz result is  
overlaid with the corresponding 4 MHz second harmonic  
and 4 MHz fundamental curves 156a, 156b and 156c  
30 respectively. The -20 dB width of the 4 MHz  
fundamental profile 156c is 6.6% narrower than the 4

1 MHz second harmonic profile 156b. The -20 dB width  
for this and multiple propagation -averaged results  
are given in Table 3 of Figure 15.

5 Each of the two-way average profiles shown  
in Figure 14a was then radially summed. Each of the  
average profiles consisted of 499 radial position-  
magnitude value pairs,  $(r_i, m_i)$ ,  $i = 1, \dots, 499$ . The 499  
pairs discretely described the averaged of 5 abdominal  
wall-jittered two-way profiles over a radial extent of  
1.2 cm. The discrete radial summation of the average  
10 profiles was defined by 499 radial position-summation  
value  $(r_j, s_j)$  pairs, where each  $s_j$  was defined by

$$s_j = \pi \sum_{i=1}^j [(r_i^2 - r_{i-1}^2) \times (m_i + m_{i-1}) / 2]$$

15 The first term in the summation involves the  $(r_o, m_o)$   
on-axis magnitude value.

The resulting radial summation or  
integration profiles are shown in Figure 14b. In  
20 Figure 14b, the integration profiles for the 2 MHz  
fundamentals, the 4 MHz second harmonics and the 4 MHz  
fundamentals are shown at 160a, 160b and 160c  
respectively. Each of these integrated two-way  
profiles was scaled such that the value at an off-axis  
25 radial distance of 1.2 cm was unity. The elevated  
sidelobes of the 2 and 4 MHz fundamental profiles  
cause their summation profiles to rise significantly  
beyond the radial extent of the mainlobe. This  
additional rise represents the potential for  
30 scattering from the sidelobes to significantly reduce  
the contrast resolution of the image. In Table 3 the

radial extent at which these integration profiles  
1 reach the 0.9 level is given. This radial extent is a  
measure of the sidelobe's potential to reduce the  
contrast resolution of an image. In this case, the  
second harmonic's radial extent is 38% less than that  
5 of it's 2 MHz fundamental and 63% less than the  
corresponding 4 MHz fundamental radial extent.

Alternatively, the summation profiles  
depicted in Figure 14b offer the percent of the two-  
way field's amplitude inside or outside a given  
10 radius. For example, 91.7% of the 2 MHz fundamental's  
amplitude, 96.7% of the 4 MHz second harmonic's  
amplitude, and 88.6% of the 4 MHz fundamental's  
amplitude fall inside a radius of 0.25 cm. The  
corresponding percentages falling outside of 0.25 cm  
15 are 8.3%, 3.3% and 11.4%, respectively. Ratios of  
these outside percentages could be useful for  
inferring the relative contrasts offered in imaging a  
low scattering or void region of a given size. For a  
void region approximately 0.5 cm across, the 4 MHz  
20 second harmonic of this device might then provide 2.5  
times (8 dB) higher contrast than the 2 MHz  
fundamental and 3.5 times (11 dB) higher contrast than  
the corresponding 4 MHz fundamental.

The previous two sets (at 2 and 4 MHz) of 5  
25 abdominal wall-jittered propagations were repeated at  
twice the source frequencies. Figures 15a and 15b  
depict the corresponding averaged results from these 4  
MHz nonlinear and 8 MHz linear propagations. In  
Figure 15a the radial average amplitude curves  
30 obtained by averaging the 5 two-way data sets are  
shown. Figure 15a shows that the 8 MHz fundamental

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mainbeam 162c is broader than the corresponding 4 MHz  
1 fundamental mainbeam 162a. The jitter-imposed lateral  
resolution limits have been encountered and in fact  
exceeded at this point. Also note that the 8 MHz  
second harmonic mainbeam 162b is narrower than either  
5 of the fundamentals 162a and 162c. Figure 15b depicts  
the additional sidelobe corruption of the fundamental  
beams and the corresponding increase in the second  
harmonic's relative contrast resolution potential.

Figures 16a, and 16b show average results  
10 from 5 propagations through breast wall delay data.  
The normalized two-way average radii for the 2 MHz  
fundamentals, the 4 MHz second harmonics and the 4 MHz  
fundamentals are shown at 166a, 166b and 166c  
respectively in Figure 16a; and the corresponding  
15 radially integrated magnitudes are shown at 170a, 170b  
and 170c respectively in Figure 16b.

In Figures 16a and 16b the results  
considered were from 2 MHz nonlinear and 4 MHz linear  
propagations. In Figure 16a, the 4 MHz mainbeams 166c  
20 are significantly broadened over the corresponding  
abdominal wall-jittered mainbeams in Figure 14a. The  
sidelobe levels in Figure 16a are also higher than  
those in Figure 14a. These Figures show that the  
breast wall layers produced significantly more  
25 distortion than the abdominal wall layers. The second  
harmonic 4 MHz profile 166b is 8.4% narrower at the  
-20 dB level than the fundamental 4 MHz 166c. In  
Figure 16b the integrated profile of the second  
harmonic 170b has a radial extent at the 0.9 level  
30 which is 48% less than the 2 MHz fundamental's 170a  
and 70% less than the 4 MHz fundamental's 170c. Thus,

in the more distorting breast wall-jittered  
1 propagations, the relative advantages of the 4 MHz  
second harmonic were larger than in the abdominal wall  
propagations and included a slightly narrower  
mainbeam. The 4 and 8 MHz results for breast wall-  
5 jittered propagations follow closely the developments  
seen in the 4 and 8 MHz curves of Figures 16a and 16b.  
Results from these propagations are included in Table  
3 of Figure 17.

10 Finally, in all of the jittered propagations  
considered, the second harmonic mainbeam was narrower  
than the fundamental mainbeam. The limits on the  
lateral resolution of the linear harmonics eventually  
put limits on the second and other higher harmonics,  
though.

15 The above discussions show the liver-path  
beam patterns for a focused Gaussian-apodized  
transducer operating at 2, 4 and 8 MHz. The non-phase  
aberrated propagations show that the second and higher  
harmonics formed through finite amplitude distortion  
20 have much lower sidelobe levels than their fundamental  
harmonic or the corresponding linear fundamentals.  
The finite amplitude production of these higher  
harmonic beams allow this sidelobe relationship to  
hold for any focused or unfocused transducer. Pulse  
25 propagation analysis shows that the higher harmonics  
formed in a propagating pulse-beam can be very well  
described by considering the harmonics produced in the  
corresponding continuous wave propagation. Modeling  
results also show that second harmonic levels  
30 sufficient for imaging purposes can be easily obtained  
within the field amplitude limits of the mechanical



index.

1           The introduction of phase jitter as computed  
from measured propagation delays from slices of  
abdominal wall and breast wall causes the sidelobe  
levels of the second harmonic and fundamental beams to  
5   rise significantly. In all of the aberrated  
propagations considered, the two-way profile of the  
second harmonic offered narrower -20 dB mainlobe  
widths and lower sidelobe levels than the fundamental  
beam which produced it. These same second harmonic  
10   profiles had slightly broader mainlobes at 4 MHz in  
abdominal wall-jitter propagations than the 4 MHz  
fundamental profiles. In all other jittered  
propagations considered, though, the second harmonic  
offered slightly narrower mainlobes than the same-  
15   frequency fundamental and substantially lower sidelobe  
levels. Thus, second harmonic-based ultrasonic images  
offer significant improvement in the lateral component  
of contrast resolution.

20           One obstacle to obtaining such images are  
artifacts from source contributions to the higher  
harmonic bandwidths. Figure 18a depicts such an  
imperfect source wave. This on-axis waveform is  
depicted in particle velocity units and corresponds to  
a peak pressure of one half of that of the pulse shown  
25   in Figure 6a. The pulse in Figure 18a has the same  
form as that in Figure 6a with the exception of the  
initial zero portion. In Figure 18b, the  
corresponding source pulse spectrum is shown.  
Significant energy content outside the 2 MHz bandwidth  
30   is visible. The resulting computed focal waveform's  
spectrum is depicted in Figure 18c and can be compared

1 to the earlier focal spectrum in Figure 6b. The log-  
scaled depiction in Figure 18c shows that a simple  
high pass filtering of the received spectrum contains  
significant contributions from the source.

5 In Figure 18d, the resulting nonlinear  
distortion pulse obtained by high pass filtering the  
spectrum of Figure 18c is shown. This pulse can be  
compared to the corresponding distortion pulse shown  
in Figure 6c. In both cases, the cut-off frequency  
was 3 MHz. In Figure 18d the full 8 microsecond  
10 period of the computed pulse is shown so that the  
ringing associated with the linear or source content  
within the second harmonic bandwidth can be seen.  
This ringing could adversely affect the axial  
component of the contrast resolution and the lateral  
15 gains associated with the lower sidelobe levels. In  
this case, improvements can be obtained by increasing  
the source amplitude so as to boost the second  
harmonic bandwidth levels.

20 A two pulse scheme may be used to alleviate  
or to eliminate such problems. In this method, two  
source pulses are sent in place of a single pulse in  
the image formation cycle. The two pulses are  
identical in form but one is significantly lower in  
amplitude. The received echo from this lower  
25 amplitude pulse is then used to remove the linear  
content from the high amplitude pulse. This is  
accomplished by subtracting an appropriately-scaled  
version of the received low amplitude signal from the  
corresponding high amplitude signal. The resulting  
30 difference signal may then be high-pass filtered  
followed by the normal sequence of image formation

steps. The high-pass filtering is preferred since  
1 pulse analysis has revealed that the low frequency  
content of the difference signal is radially wide-  
spread and reduces the higher harmonic sidelobe  
advantage.

5           Figures 19a, 19b and 19c show results  
relevant to an implementation of the two pulse scheme  
using the pulse shown in Figure 19a as the high  
amplitude source pulse. In Figure 19a, the focal  
spectrum from Figure 18c is shown at 176a and is  
10 overlaid with the focal spectrum 176b produced by a  
half-amplitude version of the same source. The low or  
half amplitude spectrum was multiplied by two and  
subtracted from the high amplitude spectrum to obtain  
the difference spectrum depicted in Figure 19b. Note  
15 that if the propagations had not involved finite  
amplitude distortion, then this difference spectrum  
would have been all zero. The difference spectrum was  
then high pass filtered and inverse transformed to  
obtain the effective on-axis distortion imaging pulse  
20 shown in Figure 19c. The transition for the high pass  
filtering was at 2.75 MHz. This same frequency was  
also appropriate for filtering off-axis difference  
spectra.

25           This two pulse scheme appears to be capable  
of extracting the desired largely-second harmonic  
images from any realistic ultrasonic imaging pulse.  
The third harmonic bandwidth depicted in Figure 19b,  
though, does not appear to be separable from the  
second harmonic bandwidth. The nodal depth between  
30 these harmonics is not deep enough. Thus, third  
harmonic (or largely-third harmonic) images do not

appear to be easily obtainable with this two pulse  
1 scheme. Also, a two pulse scheme with a  $1/4000$  second  
interval between the respective high and low amplitude  
source pulses, does not have significant artifacts due  
to tissue or transducer motion.

5 The use of the second harmonic (plus a small  
contribution from higher harmonics) to form an image  
is an independent alternative to the phase correction-  
based schemes which have been and are being examined  
by other investigators as a means for improving the  
10 contrast performance of biomedical ultrasonic imaging.

The higher harmonics also offer additional  
opportunities for correcting for beam distortion. The  
amplitude of the third harmonics in the focal region  
is strongly affected by the fundamental's amplitude.  
15 As discussed above, the nonlinear production of the  
third harmonic is proportional to the product of the  
amplitudes of the fundamental and second harmonics.  
This means that much of the third harmonic's  
production occurs in the focal region where beam  
20 distortion can reduce the amplitude of the fundamental  
and second harmonic. Thus, an iterative scheme may be  
used to correct for beam defocusing using the  
amplitude of the received third harmonic for feedback.

Second harmonic images also provide for a  
25 means of reducing speckle. In particular, an image  
formed as the sum of a second harmonic image and the  
corresponding fundamental image would have less  
speckle than either of the constituent images. Since  
the second harmonic is twice the frequency of the  
30 fundamental and has a largely constant phase  
relationship with the fundamental in the mainlobe, the

1 second harmonics image's speckle pattern would be  
conveniently out of phase with that of the  
fundamental.

5 While it is apparent that the invention  
herein disclosed is well calculated to fulfill the  
objects previously stated, it will be appreciated that  
numerous modifications and embodiments may be devised  
by those skilled in the art, and it is intended that  
the appended claims cover all such modifications and  
embodiments as fall within the true spirit and scope  
10 of the present invention.

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